The modern computer tomography (CT) an included variety of measuring modalities. On the one hand the conventional CT exists, thus the receptacle of individual layers. This became to a large extent by the volume recording technology spiral CT replaced. On the other hand recently CT-apparatuses become made, which possess several lines (N lines) instead of only one detector line. With this Mehrzeilencomputertomographen both tomographies from now N are layers simultaneous, and spiral photographs possible.

All these modalities common that a convention ("z-interpolation") exists X, the measured data p are (beta, alpha, n) (beta is the angle in compartments, alpha of the Projektionswinkel thus the angular position of the tubes and/or. Angular position of tubes and detector system and n counts the number of the detector line or the number of the tomography) on a planar transaxialen data set valid at the z-position zR pX to convert (beta, alpha, zR) (layer data record). This layer data record becomes then reconstructed with a reconstruction procedure for planar data (usually filtered Rückprojektion or Fourierrekonstruktion), around the CT-image obtained.

With conventional receptacles (tomographies), those from single circulation at positions z', z ", z '". . . develops, becomes usually however not necessarily the reconstruction only at accurate these positions allowed (D. h. zR ELEMENT z', z ", z '". . .). With spiral photographs the reconstruction position is more selectable zR generally free and retrospective. The furthest common convention to the z-interpolation with Einzeilen spiral CT is the algorithm X = 180 DEG left, a linear interpolation between data points measured in z-direction [Willi A. Calendar Wolfgang Seissler, serious block and Peter Vock, "spiral volumetric CT with single breath hold technique, continuous transport, and continuous scanner rotation", Radiology 176 (1), S. 181-183, July 1990], [Arkadiusz Polacin, Willi A. Calendar and Guy Marchal, "evaluation OF section sensitivity profile and image noise in spiral CT", Radiology 185 (1), S. 29-35, October 1992].

With all CT-images it applies that the picture quality rises regarding noises and low contrast recognizability monotonic with the patient dose. That is, that the user the tube stream and thus the dose rate of the patient so far increased, until the picture quality is its (subjective) feeling after "good". Views of dose have in particular in European countries an high value and many trials are thus undertaken to reduce the patient dose with as constant a picture quality as possible.

A simple possibility for the reduction of the pixel noise is offered by the choice of a smoothing reconstruction filter. The reconstruction filters can become at each kommeriziell erwerblichen CT-apparatus in certain limits free selected and thus can each user the intoxication level in the image reduce, without increasing the patient dose. Since becomes smoothed with this method the entire data set, this accompanies inevitably with a deterioration of the local dissolution. The problem to reduce the patient dose with as constant a picture quality as possible and/or. the picture quality with same patient dose to increase is not dissolved thereby, because it must become a compromise between local dissolution and pixel noise made.

To the solve the problem approaches are to the adaptive filtration of the measurement data, D in the literature. h. the data set becomes not global, but only local smoothed [Jiang Hsieh. "Adaptive ones trimmed mean for computed tomography image reconstruction", Proc. OF SPIE, 2299, S. 316-324, 1994], [Jiang Hsieh, "Generalized adaptive median filter and their application in computed tomography", Proc. OF SPIE, 2298, S. 662-672, 1994], [Jiang Hsieh, "adaptive one felt ring approach ton the streaking artifact reduction due tons x-ray photon starvation", Radiology 205 (P), S. 391, 1997], [Berkman Sahiner and Andrew E. Yagle, "Reconstruction from projections under time frequencies constraints", IEEE Transactions on Medical Imaging, 14 (2), S. 193-204, June 1995]. Usually adjacent detector elements become the used filtration take place thus exclusive in beta - direction the adaptive filtration.

Approaches to accomplish the filtration raw data-based adaptive (matched at each single measured projection value) in all three dimensions (beta -, alpha - and z-direction) are so far not known.

The invention indicated in claim 1 is the basis the object, for a Computertomographen (in or Mehrzeilersystem with or without helical scanning) the picture noise and the disturbing intoxication structures, which result from correlation of adjacent pixels, as far as possible to reduce. Thus an improvement of the picture quality becomes achieved with constant patient dose and/or. a dose reduction with constant picture quality possible.

This object is dissolved according to invention by the features of the claim.

The invention is subsequent explained.

It designates pX (xi, theta, zR) the planar projection data (z. B. Attenuation values) of a CT-Scans, which is valid at the z-position zR. Xi the channel index is and corresponds in fan geometry to the angle in compartments and in parallel geometry the distance of the beam to the center of rotation. theta the projection index is enspricht and both in fan and in parallel jet geometry of the angular position of the tubes and/or. the angular position of tubes and detector. X is the name of the convention to the z-interpolation, generated with which these planar data became from the measurement. The

reconstruction position zR is retrospectively on one the measurement corresponding interval [zmin, zmax] free more selectable. In addition three examples. Example 1: In case of the Einzeiler spiral CT X = 180 is DEG left a conceivable z-Interpolationsalgorithmus, which can come here to the application. Example 2: With Mehrzeiler Spiraldaten, analogue to example 1, an z-interpolation can be the basis for standard. Example 3: With conventional CT (tomographies) both with in and with Mehrzeilern and convention X a next neighbour interpolation or a linear interpolation of two adjacent, measured layers used can become.

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With the here described reconstruction procedure for the computer tomography the planar projection data become pX (xi, theta, zR) filtered before the reconstruction adaptive according to the subsequent formula:
pAF (xi, theta, z) = INTEGRAL ONE D xi "D theta "dz" NLG DELTA xi (xi - xi") NLG DELTA theta (theta - theta ") NLG DELTA z (z - z") pX (xi ", theta", z').

Designate

- pX (xi, theta, z) the projection data in parallel or fan jet geometry, standing before feedthrough of the adaptive filtration (AF) for the order. These are calculated by the measurement data by the convention X NOTE AGONY AF for arbitrary z.
- pAF (xi, theta, z) the projection data in parallel or fan jet geometry, standing after feedthrough of the adaptive filtration for the order. These data can then the reconstruction (z. B. filtered Rückprojektion) supplied become.
- DELTA xi, DELTA theta and DELTA z the filter widths in the three coordinate directions. The filter widths are functions of the current adaptive projection value which can be filtered
pX (xi, theta, z): DELTA xi = DELTA xi (pX (xi, theta, z)), DELTA theta = DELTA theta (pX (xi, theta, z)) and DELTA z = DELTA z (pX (xi, theta, z)).
- NLG DELTA xi (.), NLG DELTA theta (.) and NLG DELTA z (.) the filter functions (axle-symmetrical with values >/= 0 and total area 1) for the smoothing in the respective coordinates. The filter widths DELTA xi, DELTA theta and DELTA z stand in each case for half widths or another characteristic wide ASS of the filter functions. Or the several widths if a same zero becomes, so reduced the filter function to a Dirac delta function and it does not take place itself in the corresponding coordinates filtration.

In a possible embodiment the choice of the filter widths with the formula could

EMI4.1
take place. Designate

- xi, theta and z the scanning distance in the three coordinate smelling: xi is the width of a detector element, theta corresponds to the detector average time and z of the half width of the layer sensitivity profile.
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x (< xi ><, >< theta ><, z)> the relative variance of the projection value pX (xi, theta, z) as measure for the intoxication level of the point (xi, theta, z).
- T< 2> the threshold value starting from that the filter widths DELTA xi, DELTA theta and DELTA z (positive) a value various of zero accept.
- C the standardisation-constant, which becomes a so selected that on falling below the threshold value T the filter widths DELTA xi, DELTA theta and DELTA z the value zero to assume. Therefore cT 2< applies> = xi theta z.

The condition (DELTA xi + xi): (DELTA theta + theta): (DELTA z + z) = xi: theta: z places safer that the ratio of the effective filter widths (these are the filter widths plus the intrinsic averaging by the CT-apparatus) corresponds to the equipmentconditional and thus ratio from detector-wide, integration time and z-expansion, optimized of the manufacturer.

The filter function knows z. B. one of the subsequent representations have:

Rectangle
EMI5.1

Triangle
EMI5.2

Gauss
EMI5.3

These representations have the advantage that the associated integrations analytic performed to become to be able and the adaptive filtration itself on a weighted sum over the measured data reduced. Thus the adaptive filters can be implemented efficiently.